ON THE EXPERIMENTAL RESULTS OF A NONINVASIVE ESTIMATION TECHNIQUE OF MUSCLE CONDUCTION VELOCITY DISTRIBUTION

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Abstract – **This paper is intended to present an assessment of the experimental results of a previously proposed muscle conduction velocity distribution (CVD) estimator. The performance of the proposed technique was seriously deteriorated when applied to experimental data. The causes for this decline were evaluated by introducing real-world errors in the model parameters and looking at how sensitive the estimator is to these. The simulation results show the high sensitivity of the estimator to parameter errors. The similitude found between the simulation results and the CVD estimates obtained on experimental data helped confirm this observation as well. Given the sensitivity observed the proposed technique is not practical for muscle dimensions such as those found in the biceps brachii.**

***Keywords* – Conduction, propagation, velocity, CVD, EMG, noninvasive, estimation.**

I. INTRODUCTION

Noninvasive techniques for the estimation of conduction velocity distribution (CVD) on human muscles can lead to new possibilities for clinical assessment of muscular pathologies. Efforts are currently directed to the development of such tools [1][2]. However due to the constraints imposed by the low frequency content of the electromyographic (EMG) signal together with physiological issues related to muscle geometry this task is a challenging one. This is true particularly when the aim is to characterize a representative part of the muscle motor unit population through a CVD estimate.

II. METHOD

The technique used by the authors is a deconvolution approach based on volume conductor modeling of the surface EMG interference pattern [2]. It makes use of two correlation functions computed from the surface EMG data recorded at the skin surface with a bipolar configuration. A volume conductor model of the signal that depends on different parameters is built into the estimator. Some of these parameters such as interelectrode spacing, which is a rather accurate quantity, are determined by the recording configuration. However, other parameters such as

1. Location of the recording electrodes with respect to the innervation zone, and
2. Muscle geometry and thickness of the fat layer under each recording channel,

are not known with certainty.

The setup shown in Figure 1 was used for experiments and simulations. Two bipolar channels spaced by a distance d_{ch} were aligned in the direction of the muscle fibers. As part of the tests implemented through simulations, the distance from the recording electrode arrangement to the middle of the innervation zone was underestimated by 5, 10 and 20mm during the CVD estimation. The power of the signals recorded from channels X & Y can also differ from that expected according to the assumption made on model parameters mentioned in point 2. These differences in channel powers convey differences in the relative amplitudes of the MES auto and cross correlation functions R_{XX} and R_{XY}, respectively. The effect that this has on the estimator performance was evaluated by upsetting the auto-to-cross amplitude ratio of the simulated signals by 5, 10 and 20%.

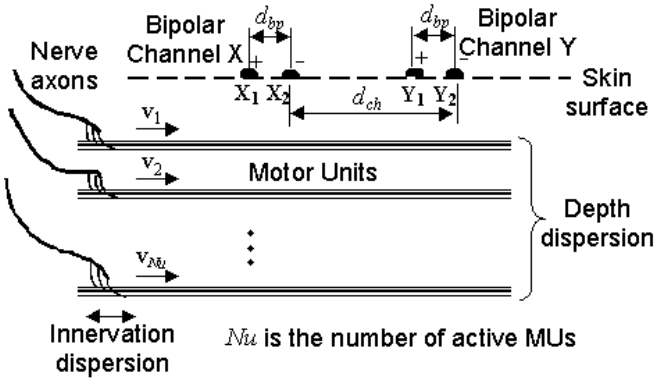

Figure 1. Disposition of two bipolar recording channels

The estimator performance was measured by means of the average estimate error per bin which is given by

$$MAE_{bin} = \frac{\frac{1}{N_b}\sum_{i=1}^{N_b}|\hat{m}_i - m_i|}{\frac{1}{N_b}\sum_{i=1}^{N_b} m_i} \times 100\ \% = \sum_{i=1}^{N_b}|\hat{m}_i - m_i| \times 100\ \%$$

where $\hat{m}_i$ is the estimate of the relative number of active fibers corresponding to velocity bin i, and m_i is the actual relative number of active fibers belonging to velocity bin i.

III. SIMULATIONS

Two channels, X & Y, using bipolar recording configurations as shown in Figure 1 were simulated. The auto correlation function R_{XX} and the cross correlation function R_{XY} were generated and fed to the estimator. Table 1 shows the MAE_{bin} performance index for different combinations of distance errors and auto-to-cross amplitude ratio errors. The channel spacing used was d_{ch} = 39.5mm. The first row presents the values obtained in the absence of errors.

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Abstract

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Error in the Location of the Innervation Zone	Error in the Relative Amplitude of Auto/Cross Corr Fns	Average Estimate Error / Bin (in % of the Average Number of Active MUs per Bin)	
		Initial Estimate	Filtered Estimate
-	-	70.6 %	13.6 %
10 mm	-	125.4 %	86.2 %
20 mm	-	109.2 %	95.8 %
10 mm	5 %	119.1 %	92.2 %
10 mm	10 %	123.5 %	99.6 %
10 mm	20 %	108.7 %	105.3 %

Table 1. Average CVD estimate errors per bin. Results correspond to estimates obtained introducing errors in the location of the innervation zone and/or errors in the relative power of the two channels. A channel spacing d_{ch} = 39.5mm was used. Estimates were obtained over 16 velocity bins.

As an error is introduced the filtered estimate index goes up from 13.6% in the first row to the 85-105% range. Figure 2 shows the CVD estimates obtained when a 10mm distance error is combined with a 5% amplitude ratio error. The estimated distributions present three significant components around 3.4m/s, 4.0-4.5m/s and 2.0m/s which is significantly large (see Figure 2) and is not present when there is no error in the estimator parameters (see Figure 3).

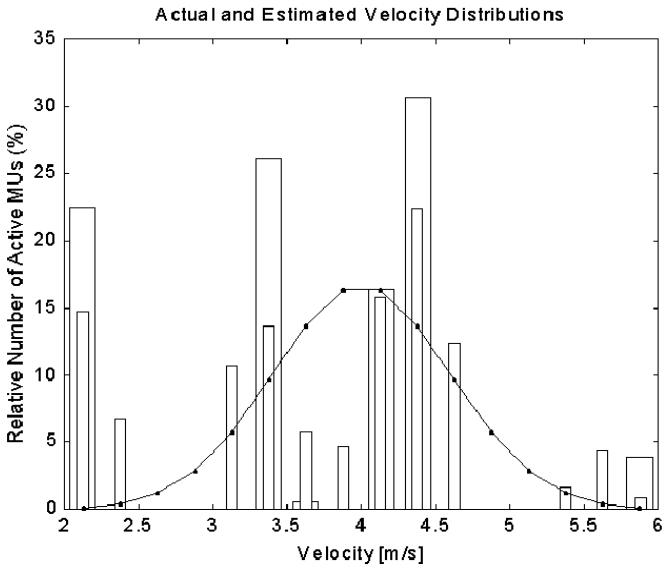

Figure 2. Primary (wide bars) and filtered (narrow bars) CVD estimates. Actual CVD used in the signal generation (curve). The distance from recording electrodes to innervation zone is underestimated by 10mm. The ratio of auto-to-cross correlation functions was upset by 5%. A channel spacing d_{ch} = 39.5mm was used.

Figure 2 shows the CVD estimate corresponding to the 4th row in Table 1. Figure 3 shows the CVD estimate corresponding to the 1st row in Table 1, that is, the parameter error free estimate. Two representations of the CVD estimate are shown on both figures in correspondence with columns 3 & 4 of Table 1. It has been shown that the smoothed or low-pass filtered estimate help decrease the variance of the estimation [2]. It can also be appreciated that the estimator error is smaller when the filtered estimate is used. Nonetheless, the detriment suffered by the estimates as a result of misjudging the location of the innervation zone by 10 mm is significant. The result is more degraded if parameters affecting the signal power in as little as 5% are misjudged as well.

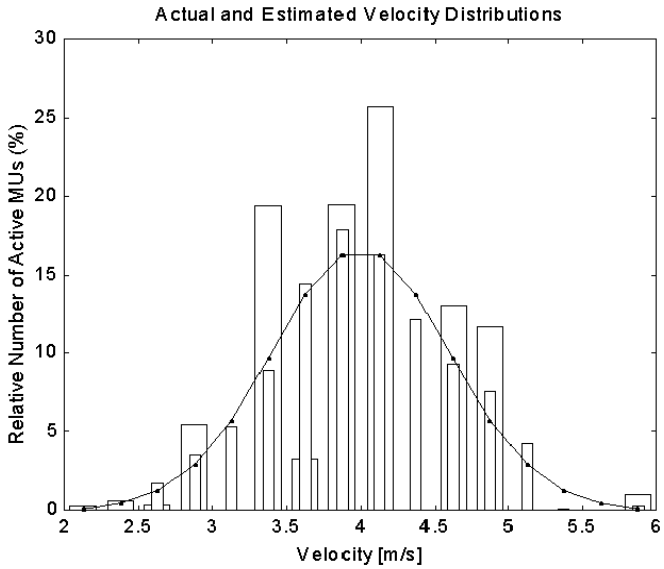

Figure 3. CVD Estimates (bars) obtained for a parameter error free estimator. The actual CVD used in the signal generation is represented by the curve. The wide bars represent the initial estimate while the narrow bars represent the filtered estimate. A channel spacing d_{ch} = 39.5mm was used.

IV. Experimental Results

Two channels of surface EMG data were collected from the biceps brachii during 1 minute of an isometric isotonic contraction. The arrangement was located on one side of the innervation zone and its axis was aligned parallel to the muscle fibers as illustrated in Figure 1. A typical CVD estimate obtained using d_{ch} = 39.5mm is shown in Figure 4. It again shows relative wide gaps in the estimated distribution. Only three significant components are present in the CVD estimate. The middle one around 4m/s corresponds to the value for the mean conduction velocity. The estimate also displays two components located at the boundaries of the estimation interval.

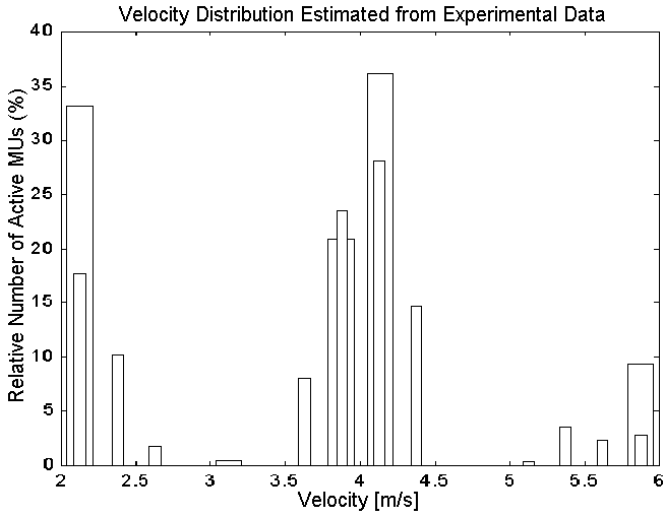

Figure 4. Primary (wide bars) and filtered (narrow bars) CVD estimates obtained from experimental EMG data collected using d_{ch} = 39.5mm.

It was observed as part of this study that the CVD estimator becomes less sensitive to errors in the model parameter as channel spacing and signal bandwidth are increased. The components shown in Figure 4 are sufficient for the algorithm to converge in the case of the posed problem. However, if a larger component population is to be obtained for the CVD estimate through this method, characteristics such as signal bandwidth and channel spacing need to be significantly increased. Since increasing these two elements are restricted by surface recording and muscle anatomy, respectively, this represents a significant obstacle to surmount when applying this type of noninvasive technique to human muscles.

V. CONCLUSION

Since the range of errors used in the simulations is very likely to be present in practice it is not surprising that the CVD estimate obtained from experimental data is seriously affected. It is apparent that the velocity component estimated at 2m/s in Figure 4 is erroneous. Such a spread in the disposition of the estimated velocity components, which suffices for the solution technique to converge, is not adequate since the velocity resolution obtained is extremely poor. One way to improve this resolution problem would be to considerably increase the channel spacing. However, this is not possible for the dimensions found in the human biceps brachii muscle if the arrangement is to be kept in one side of the innervation zone.

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